(54) Title: MULTI-PEAK SPEECH PROCESSOR

(57) Abstract

An improved pulsatile system for a cochlear prosthesis is disclosed. The system employs a multi-spectral peak codi strategy to extract a number, for example five, of spectral peaks from an incoming acoustic signal received by a microphone. encodes this information into sequential pulses that are sent to selected electrodes of a cochlear implant. The first formant (F spectral peak (280-1000 Hz) and the second formant (F2) spectral peak (800-4000 Hz) are encoded and presented to apical a basal electrodes, respectively, F1 and F2 electrode selection follows the tonotopic organization of the cochlea. High-frequen spectral information is sent to more basal electrodes and low-frequency spectral information is sent to more apical electrod Spectral energy in the regions of 2000-2800 Hz, 2800-4000 Hz, and above 4000 Hz is encoded and presented to three fixed el trodes. The fundamental or voicing frequency (F0) determines the pulse rate of the stimulation during voiced periods and a ps do-random aperiodic rate determines the pulse rate of stimulation during unvoiced periods. The amplitude of the acoustic sig in the five bands determines the stimulus intensity.

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MULTI-PEAK SPEECH PROCESSOR

Technical Field

This invention relates to pulsatile type multi-channel cochlear implant systems for the totally or profoundly deaf.

Background of the Invention

Pulsatile multi-channel cochlear implant systems generally include a cochlear implant, an external speech processor, and an external headset. The cochlear implant delivers electrical stimulation pulses to an electrode array (e.g., 22 electrodes) placed in the cochlea. The speech processor and headset transmit information and power to the cochlear implant.

The speech processor operates by receiving an incoming acoustic signal from a microphone in the headset, or from an alternative source, and extracting from this signal specific acoustic parameters. Those acoustic parameters are used to determine electrical stimulation parameters, which are encoded and transmitted to the cochlear implant via a transmitting coil in the headset, and a receiving coil forming part of the implant.

In many people who are profoundly deaf, the reason for deafness is absence of, or destruction of, the hair cells in the cochlea which transduce acoustic signals into nerve impulses. These people are thus unable to derive any benefit from conventional hearing aid systems, no matter how loud the acoustic stimulus is made, because it is not possible for nerve impulses to be generated from sound in the normal manner. Cochlear implant systems seeks to bypass these hair cells by presenting electrical stimulation to the auditory nerve fibers directly, leading to the perception of sound in the brain. There have been many ways

described in the past for achieving this object, running from implantation of electrodes in the cochlea connected to the outside world via a cable and connector attached to the patient's skull, to sophisticated multichannel devices communicating with an external computer via radio frequency power and data links.

The invention described herein is particularly suited for use in a prosthesis which comprises a multichannel electrode implanted into the cochlea, connected to a multichannel implanted stimulator unit, which receives power and data from an externally powered wearable speech processor, wherein the speech processing strategy is based on known psychophysical phenomenon, and is customized to each individual patient by use of a diagnostic and programming unit. One example of such a prosthesis is the one shown and described in U.S. Patent No. 4,532,930 to Crosby et al., entitled "Cochlear Implant System for an Auditory Prosthesis."

In order to best understand the invention it is necessary to be aware of some of the physiology and anatomy of human hearing, and to have a knowledge of the characteristics of the speech signal. In addition, since the hearing sensations elicited by electrical stimulation are different from those produced by acoustic stimulation in a normal hearing person, it is necessary to discuss the psychophysics of electrical stimulation of the auditory system. In a normal hearing person, sound impinges on the ear drum, as illustrated in FIG. 1, and is transmitted via a system of bones called the ossicles, which act as levers to provide amplification and acoustic impedance matching to a piston, or membrane, called the oval window, which is coupled to the cochlea chamber.

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The cochlear chamber is about 35 mm long
when unrolled and is divided along most of its length
by a partition. This partition is called the basilar
membrane. The lower chamber is called the scala tym-
pani. An opening at the remote end of the cochlea
chamber communicates between the upper and lower
halves thereof. The cochlea is filled with a fluid
having a viscosity of about twice that of water. The
scala tympani is provided with another piston or mem-
brane called the round window which serves to take up
the displacement of the fluid.

When the oval window is acoustically driven
via the ossicles, the basilar membrane is displaced by
the movement of fluid in the cochlea. By the nature
of its mechanical properties, the basilar membrane
vibrates maximally at the remote end or apex of the
cochlea for low frequencies, and near the base or oval
window thereof for high frequencies. The displacement
of the basilar membrane stimulates a collection of
cells called the hair cells situated in a special
structure on the basilar membrane. Movements of these
hairs produce electrical discharges in fibers of the
VIIIth nerve, or auditory nerve. Thus the nerve fi-
bers from hair cells closest to the round window (the
basal end of the cochlea) convey information about
high frequency sound, and fibers more apical convey
information about low frequency sound. This is re-
ferred to as the tonotopic organization of nerve fi-
bers in the cochlea.

Hearing loss may be due to many causes, and
is generally of two types. Conductive hearing loss
occurs when the normal mechanical pathways for sound
to reach the hair cells in the cochlea are impeded,
for example by damage to the ossicles. Conductive
hearing loss may often be helped by use of hearing

aids, which amplify sound so that acoustic information
does reach the cochlea. Some types of conductive
hearing loss are also amenable to alleviation by sur-
gical procedures.

Sensorineural hearing loss results from
damage to the hair cells or nerve fibers in the coch-
lea. For this type of patient, conventional hearing
aids will offer no improvement because the mechanisms
for transducing sound energy into nerve impulses have
been damaged. It is by directly stimulating the audi-
tory nerve that this loss of function can be partially
restored.

In the system described herein, and in some
other cochlear implant systems in the prior art, the
stimulating electrodes are surgically placed in the
scala tympani, in close proximity to the basilar mem-
brane, and currents that are passed between the elec-
trodes result in neural stimulation in groups of nerve
fibers.

The human speech production system consists
of a number of resonant cavities, the oral and the
nasal cavities, which may be excited by air passing
through the glottis or vocal cords, causing them to
vibrate. The rate of vibration is heard as the pitch
of the speaker's voice and varies between about 100
and 400 Hz. The pitch of female speakers is generally
higher than that of male speakers.

It is the pitch of the human voice which
gives a sentence intonation, enabling the listener,
for instance, to be able to distinguish between a
statement and a question, segregate the sentences in
continuous discourse and detect which parts are par-
ticularly stressed. This together with the amplitude
of the signal provides the so-called prosodic informa-
tion.

Speech is produced by the speaker exciting the vocal cords, and manipulating the acoustic cavities by movement of the tongue, lips and jaw to produce different sounds. Some sounds are produced with the vocal cords excited, and these are called voiced sounds. Other sounds are produced by other means, such as the passage of air between teeth and tongue, to produce unvoiced sounds. Thus the sound "Z" is a voiced sound, whereas "S" is an unvoiced sound; "B" is a voiced sound, and "P" is an unvoiced sound, etc.

The speech signal can be analyzed in several ways. One useful analysis technique is spectral analysis, whereby the speech signal is analyzed in the frequency domain, and a spectrum is considered of amplitude (and phase) versus frequency. When the cavities of the speech production system are excited, a number of spectral peaks are produced, and the frequencies and relative amplitudes of these spectral peaks are also varied with time.

The number of spectral peaks ranges between about three and five and these peaks are called "formants". These formants are numbered from the lowest frequency formant, conventially called F1, to the highest frequency formants, and the voice pitch is conventionally referred to as F0. Characteristic sounds of different vowels are produced by the speaker changing the shape of the oral and nasal cavities, which has the effect of changing the frequencies and relative intensities of these formants.

In particular, it has been found that the second formant (F2) is important for conveying vowel information. For example, the vowel sounds "oo" and "ee" may be produced with identical voicings of the vocal cords, but will sound different due to different second formant characteristics.

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There is of course a variety of different
sounds in speech and their method of production is
complex. For the purpose of understanding the inven-
tion herein however, it is sufficient to remember that
there are two main types of sounds - voiced and
unvoiced; and that the time course of the frequencies
and amplitudes of the formants carries most of the
intelligibility of the speech signal.

The term "psychophysics" is used herein to
refer to the study of the perceptions elicited in
patient's by electrical stimulation of the auditory
nerve. For stimulation at rates between 100 and 400
pulses per second, a noise is perceived which changes
pitch with stimulation rate. This is such a distinct
sensation that it is possible to convey a melody to a
patient by its variation.

By stimulating the electrode at a rate pro-
portional to voice pitch (F0), it is possible to con-
vey prosodic information to the patient. This idea is
used by some cochlear implant systems as the sole
method of information transmission, and may be per-
formed with a single electrode.

It is more important to convey formant in-
formation to the patient, as this contains most of the
intelligibility of the speech signal. It has been
discovered by psychophysical testing that just as an
auditory signal which stimulates the remote end of the
cochlea produces a low frequency sensation and a sig-
nal which stimulates the near end thereof produces a
high frequency sensation, a similar phenomenon will be
observed with electrical stimulation. The perceptions
elicited by electrical stimulation at different posi-
tions inside the cochlea have been reported by the
subjects as producing percepts which vary in "sharp-
ness" or "dullness", rather than pitch as such. How-

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ever, the difference in frequency perceptions between
electrodes is such that formant, or spectral peak,
information can be coded by selection of electrode, or
site of stimulation in the cochlea.

It has been found by psychophysical testing
that the perceived loudness of sounds elicited by
electrical stimulation of the auditory nerve has a
larger dynamic range then the dynamic range of the
stimulation itself. For example, a 220 dB dynamic
range of electrical stimulation may produce percep-
tions from threshold or barely perceivable, to thresh-
old of pain. In normal hearing people the dynamic
range of sound perception is in the order of 100 dB.

It has also been discovered through psycho-
physical testing that the pitch of sound perceptions
due to electrical stimulation is also dependent upon
frequency of stimulation, but the perceived pitch is
not the same as the stimulation frequency. In partic-
ular, the highest pitch able to be perceived through
the mechanism of the changing stimulation rate alone
is in the order of 1 kHz, and stimulation at rates
above this maximum level will not produce any increase
in frequency or pitch of the perceived sound. In
addition, for electrical stimulation within the coch-
lea, the perceived pitch depends upon electrode posi-
tion. In multiple electrode systems, the perceptions
due to stimulation at one electrode are not indepen-
dent of the perceptions due to simultaneous stimula-
tion of nearby electrodes. Also, the perceptual qual-
ities of pitch, "sharpness", and loudness are not
independently variable with stimulation rate, elec-
trode position, and stimulation amplitude.

Some systems of cochlear implants in the
prior art are arranged to stimulate a number of elec-
trodes simultaneously in proportion to the energy in

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specific frequency bands, but this is done without
reference to the perceptions due to stimulus current
in nearby stimulating electrodes. The result is that
there is interaction between the channels and the
loudness is affected by this.

A number of attempts have heretofore been
made to provide useful hearing through electrical
stimulation of auditory nerve fibers, using electrodes
inside or adjacent to some part of the cochlear struc-
ture. Systems using a single pair of electrodes are
shown in U.S. Patent No. 3,751,605 to Michelson and
U.S. Patent No. 3,752,939 to Bartz.

In each of these systems an external speech
processing unit converts the acoustic input into a
signal suitable for transmission through the skin to
an implanted receiver/stimulator unit. These devices
apply a continuously varying stimulus to the pair of
electrodes, stimulating at least part of the popula-
tion of auditory nerve fibers, and thus producing a
hearing sensation.

The stimulus signal generating from a given
acoustic input is different for each of these systems,
and while some degree of effectiveness has been demon-
strated for each, performance has varied widely across
systems and also for each system between patients.
Because the design of these systems has evolved empir-
ically, and has not been based on detailed psychophys-
ical observations, it has not been possible to deter-
mine the cause of this variability. Consequently, it
has not been possible to reduce it.

An alternative approach has been to utilize
the tonotopic organization of the cochlea to stimulate
groups of nerve fibers, depending on the frequency
spectrum of the acoustic signal. Systems using this
technique are shown in U.S. Patent No. 4,207,441 to

Ricard, U.S. Patent No. 3,449,753 to Doyle, U.S. Patent No. 4,063,048 to Kissiah, and U.S. Patents No. 4,284,856 and No. 4,357,497 to Hochmair et al.

The system described by Kissiah uses a set of analog filters to separate the acoustic signal into a number of frequency components, each having a predetermined frequency range within the audio spectrum. These analog signals are converted into digital pulse signals having a pulse rate equal the frequency of the analog signal they represent, and the digital signals are used to stimulate the portion of the auditory nerve normally carrying the information in the same frequency range. Stimulation is accomplished by placing an array of spaced electrodes inside the cochlea.

The Kissiah system utilizes electrical stimulation at rates up to the limit of normal acoustic frequency range, say 10 kHz, and independent operation of each electrode. Since the maximum rate of firing of any nerve fiber is limited by physiological mechanisms to one or two kHz, and there is little perceptual difference for electrical pulse rates above 800 Hz, it may be inappropriate to stimulate at the rates suggested. No consideration is given to the interaction between the stimulus currents generated by different electrodes, which experience shows may cause considerable uncontrolled loudness variations, depending on the relative timing of stimulus presentations. Also, this system incorporates a percutaneous connector which has with it the associated risk of infection.

The system proposed by Doyle limits the stimulation rate for any group of fibers to a rate which would allow any fiber to respond to sequential stimuli. It utilizes a plurality of transmission channels, with each channel sending a simple composite

power/data signal to a bipolar pair of electrodes. Voltage source stimulation is used in a time multiplexed fashion similar to that subsequently used by Ricard and described below, and similar uncontrolled loudness variations will occur with the suggested independent stimulation of neighboring pairs of electrodes. Further, the requirement of a number of transmission links equal to the number of electrode pairs prohibits the use of this type of system for more than a few electrodes.

The system proposed by Ricard utilizes a filter bank to analyze the acoustic signal, and a single radio link to transfer both power and data to the implanted receiver/stimulator, which presents a time-multiplexed output to sets of electrodes implanted in the cochlea. Monophasic voltage stimuli are used, with one electrode at a time being connected to a voltage source while the rest are connected to a common ground line. An attempt is made to isolate stimulus currents from one another by placing small pieces of silastic inside the scala, between electrodes. Since monophasic voltage stimuli are used, and the electrodes are returned to the common reference level after presentation of each stimulus, the capacitive nature of the electrode/electrolyte interface will cause some current to flow for a few hundred microseconds after the driving voltage has been returned to zero. This will reduce the net transfer of charge (and thus electrode corrosion) but this charge recovery phase is now temporarily overlapped with the following stimulus or stimuli. Any spatial overlap of these stimuli would then cause uncontrolled loudness variations.

In the Hochmair et al. patents a plurality of carrier signals are modulated by pulses correspond-

ing to signals in audio frequency bands. The carrier
signals are transmitted to a receiver having indepen-
dent channels for receiving and demodulating the
transmitted signals. The detected pulses are applied
to electrodes on a cochlear implant, with the elec-
trodes selectively positioned in the cochlea to stimu-
late regions having a desired frequency response. The
pulses have a frequency which corresponds to the fre-
quency of signals in an audio band and a pulse width
which corresponds to the amplitude of signals in the
audio band.

U.S. Patent No. 4,267,410 to Forster et al.
describes a system which utilizes biphasic current
stimuli of predetermined duration, providing a good
temporal control of both stimulating and recovery
phases. However, the use of fixed pulse duration
prohibits variation of this parameter which may be
required by physiological variations between patients.
Further, the data transmission system described in
this system severely limits the number of pulse rates
available for constant rate stimulation.

U.S. Patent No. 4,593,696 to Hochmair et al.
describes a system in which at least one analog elec-
trical signal is applied to implanted electrodes in a
patient, and at least one pulsatile signal is applied
to implanted electrodes. The analog signal represents
a speech signal, and the pulsatile signal provides
specific speech features such as formant frequency and
pitch frequency.

U.S. Patent No. 4,515,158 to Patrick et al.
describes a system in which sets of electrical cur-
rents are applied to selected electrodes in an im-
planted electrode array. An incoming speech signal is
processed to generate an electrical input correspond-
ing to the received speech signal, and electrical

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signals characterizing accoustic features of the
speech signal are generated from the input signal.
Programmable means obtains and stores data from the
electrical signals and establishes sets of electric
stimuli to be applied to the electrode array, and
instruction signals are produced for controlling the
sequential application of pulse stimuli to the elec-
trodes at a rate derived from the voicing frequency of
the speech signal for voiced utterances and at a rate
independent of the voicing frequency for unvoiced
utterances.

      The state of the art over which the present
invention represents an improvement is perhaps best
exemplified by the aforesaid U.S. Patent No. 4,532,930
to Crosby et al., entitled "Cochlear Implant System
for an Auditory Prosthesis". The subject matter of
said Crosby et al. patent is hereby incorporated here-
in by reference. The Crosby et al. patent describes a
cochlear implant system in which an electrode array
comprising multiple platinum ring electrodes in a
silastic carrier is implanted in the cochlea of the
ear. The electrode array is connected to a multi-
channel receiver-stimulating unit, containing a semi-
conductor integrated circuit and other components,
which is implanted in the patient adjacent the ear.
The receiver-stimulator unit receives data information
and power through a tuned coil via an inductive link
with a patient-wearable external speech processor.
The speech processor includes an integrated circuit
and various components which are configured or mapped
to emit data signals from an Erasable Programmable
Read Only Memory (EPROM). The EPROM is programmed to
suit each patient's electrical stimulation percep-
tions, which are determined through testing of the
patient and his implanted stimulator/electrode. The

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testing is performed using a diagnostic and program-
ming unit (DPU) that is connected to the speech pro-
cessor by an interface unit.

The Crosby et al. system allows use of
various speech processing strategies, including domi-
nant spectral peak and amplitude compression of voice
pitch, so as to include voiced sounds, unvoiced glot-
tal sounds and prosodic information. The speech pro-
cessing strategy employed is based on known psycho-
physical phenomenon, and is customized to each indi-
vidual patient by the use of the diagnostic and pro-
gramming unit. Biphasic pulses are supplied to vari-
ous combinations of the electrodes by a switch con-
trolled current sink in various modes of operation.
Transmission of data is by a series of discrete data
bursts which represent the chosen electrode(s), the
electrode mode configuration, the stimulating current,
and amplitude determined by the duration of the ampli-
tude burst.

Each patient will have different perceptions
resulting from electrical stimulation of the cochlea.
In particular, the strength of stimulation required to
elicit auditory perceptions of the same loudness may
be different from patient to patient, and from elec-
trode to electrode for the same patient. Patients
also may differ in their abilities to perceive pitch
changes from electrode to electrode.

The speech processor accommodates differ-
ences in psychophysical perceptions between patients
and compensates for the differences between electrodes
in the same patient. Taking into account each indi-
vidual's psychophysical responses, the speech proces-
sor encodes acoustic information with respect to stim-
ulation levels, electrode frequency boundaries, and
other parameters that will evoke appropriate auditory

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perceptions. The psychophysical information used to
determine such stimulation parameters from acoustic
signals is referred to as a MAP and is stored in a
random access memory (RAM) inside the speech proces-
sor. An audiologist generates and "fine tunes" each
patient's MAP using a diagnostic and programming sys-
tem (DPS). The DPS is used to administer appropriate
tests, present controlled stimuli, and confirm and
record test results.

The multi-electrode cochlear prosthesis has
been used successfully by profoundly deaf patients for
a number of years and is a part of everyday life for
many people in various countries around the world.
The implanted part of the prosthesis has remained
relatively unchanged except for design changes, such
as those made to reduce the overall thickness of the
device and to incorporate an implanted magnet to elim-
inate the need for wire headsets.

The external speech processor has undergone
significant changes since early versions of the pros-
thesis. The speech coding scheme used by early pa-
tients presented three acoustic features of speech to
implant users. These were amplitude, presented as
current level of electrical stimulation; fundamental
frequency or voice pitch, presented as rate of pulsa-
tile stimulation; and the second formant frequency,
represented by the position of the stimulating elec-
trode pair. This coding scheme (FOF2) provided enough
information for profoundly postlinguistically deafened
adults to show substantial improvements in their per-
ception of speech.

The early coding scheme progressed naturally
to a later coding scheme in which additional spectral
information is presented. In this scheme a second
stimulating electrode pair was added, representing the

first formant of speech. The new scheme (FOF1F2) showed
improved performance for adult patients in all areas of
speech perception.

Despite success of speech processors using the FOF1F2
scheme over the last few years, a number of problems have
been identified. For example, patients who perform well in
quiet conditions can have significant problems when there is
a moderate level of background noise. Also, the FOF1F2 scheme
codes frequencies up to about 3,500Hz; however, many phonemes
and environmental sounds have a high proportion of their
energy above this range making them inaudible to the implant
user in some cases.

According to one aspect of the invention there is
provided an improved pulsatile system for a cochlear
prosthesis in which an incoming audio signal is concurrently
presented to a speech feature extractor and a plurality of
band pass filters, the pass bands of which are different from
one another and at least one of which is at a higher
frequency than the normal range of the second formant or
frequency peak of the speech signal. The energy within these
pass bands controls the amplitude of electrical stimulation
of a corresponding number of fixed electrode pairs adjacent
the basal end of the electrode array, thus providing
additional information about high frequency sounds at a
tonotopically appropriate place within the cochlea.
Preferably three additional band pass filters are employed in
the ranges of 2,000 to 2,800Hz, 2,800 to 4,000Hz and 4,000 to

8,000Hz.

The overall stimulation rate remains as FO (fundamental frequency or voice pitch) but, in addition, four electrical stimulation pulses occur for each glottal pulse, as compared with the FOF1F2 strategy heretofore used, in which only two pulses occur per voice pitch period. For voiced speech sounds, pulses representing the first and second formant are provided along with additional stimulation pulses representing energy in the 2,000 to 2,800Hz and the 2,800 to 4,000Hz ranges. For unvoiced phonemes, yet another pulse representing energy above 4,000Hz is provided while no stimulation for the first formant is provided, since there is no energy in this frequency range. Stimulation occurs at a random pulse rate of approximately 260Hz, which is about double that used in earlier speech coding schemes.

According to another aspect of the invention there is provided a method of processing an audio spectrum signal received from a microphone to produce signals for stimulating a patient implantable tissue stimulating multi-channel electrode array adapted to be positioned in a cochlea from the apical region of the cochlea to the basal region of the cochlea, said method comprising selecting a first dominant frequency peak from said audio signal from a frequency band of between about 280Hz and about 1,000Hz and stimulating at least one electrode in the apical region of said electrode array in accordance with the spectral information contained in said first peak; selecting a second dominant frequency

peak from said audio signal from a frequency band of between

about 800Hz and about 4,000Hz and stimulating at least one

electrode in the basal region of said electrode array in

accordance with the spectral information contained in said

second peak; extracting spectral information in at least one

region of the spectrum of said audio signal and stimulating

at least one predetermined electrode in said electrode array

in accordance with said extracted spectral information, said

predetermined electrode being in said basal region of said

electrode array.

Preferably, additional preselected electrodes are

stimulated using spectral energy derived from said audio

signal in the audio frequency regions 2,000 to 2,800Hz, 2,800

to 4,000Hz and above 4,000Hz respectively.

In accordance with another aspect of the invention

there is provided an improved speech processor for a cochlear

prosthesis which employs a multi-spectral peak (MPEAK) coding

strategy to extract a number, for example five, of spectral

peaks from an incoming acoustic signal received by a

microphone. The speech processor encodes this information

into sequential pulses that are sent to selected electrodes

of a cochlear implant. The first formant (F1) spectral peak
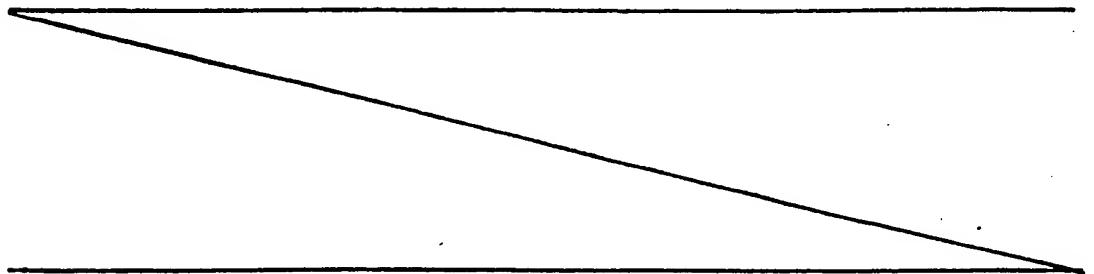
(280-1000 Hz) and the second formant (F2) spectral
peak (800-4000 Hz) are encoded and presented to apical
and basal electrodes, respectively. F1 and F2 elec-
trode selection follows the tonotopic organization of
the cochlea. High-frequency spectral information is
sent to more basal electrodes and low-frequency spec-
tral information is sent to more apical electrodes.
Spectral energy in the regions of 2000-2800 Hz, 2800-
4000 Hz, and above 4000 Hz is encoded and presented to
three fixed electrodes. The fundamental or voicing
frequency (F0) determines the pulse rate of the stimu-
lation during voiced periods and a pseudo-random ape-
riodic rate determines the pulse rate of stimulation
during unvoiced periods. The amplitude of the acous-
tic signal in the five bands determines the stimulus
intensity.

Brief Description of the Drawings

        While the specification concludes with
claims particularly pointing out and distinctly claim-
ing the subject matter of this invention, it is be-
lieved that the invention will be better understood
from the following description, taken in conjunction
with the accompanying drawing, in which:

        FIGS. 1A and 1B are interior views of the
anatomy of a human ear and a cross section of a coch-
lea, respectively;

        FIG. 2 is a block diagram of the overall
cochlea implant system of this invention;

        FIG. 3 is a pictorial view of the components
of this system, including the implantable parts and
the parts worn by the patient;

        FIGS. 3A and 3B are respective side and end
elevation views of the implantable parts of this sys-
tem;

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FIG. 4 is a graph of current vs. time, showing the biphasic current waveform utilized in this invention;

FIG. 5 is a graph showing an example of the sequential stimulation pattern of electrode pairs for a voiced sound using the multi-peak coding strategy of this invention;

FIG. 6 is a graph showing an example of the sequential stimulation pattern of electrode pairs for an unvoiced sound using the multi-peak coding strategy of this invention;

FIG. 7 is a chart showing an example of the pattern of electrical stimulation for various steady-state phonemes using the multi-peak coding strategy of this invention;

FIG. 8 is a graph showing the standard loudness growth function for the speech processor of this invention; and,

FIG. 9 is a block diagram of the microphone and speech processor portions of a pulsatile type, multi-channel cochlear implant system in accordance with this invention.

Best Mode for Carrying Out the Invention

The cochlear implant system of this invention, shown in FIG. 2, comprises several components. An electrode array 1 is implanted into the cochlea. The electrode array 1 comprises a number of rings or bands of platinum molded with a flexible silastic carrier. Preferably, there are 32 bands of platinum in total. The distal 22 bands are active electrodes, and have connecting wires welded to them. The proximal 10 electrode bands are used for stiffening, and to act as an aid to surgical insertion. In a typical array, the electrode rings are about 0.05 mm in thickness with a width of 0.3 mm, and have outside diame-

ters ranging from 0.6 mm at the proximal end to 0.4 mm diameter at the distal end. The diameter of the rings changes smoothly so that the array is tapered over the distal 10 mm or so. The rings are spaced on 0.75 mm centers over the distal 25 mm of the electrode array, and all of the exposed outside area of the rings is used as active electrode area. The silastic material may be MDX4-4210, manufactured by Dow Corning.

The 22 electrode wires pass via a cable 2 from the electrode array 1 to the receiver-stimulator unit (RSU) 3. The invention described is not limited to the use of this design of electrode array, and a number of alternative electrode designs as have been described in the prior art could be used. The RSU 3 receives information and power from an external source through a tuned receiving coil 5 attached to the RSU and positioned just beneath the skin. The RSU also provides electrical stimulating pulses to the electrode array 1. The power, and data on which electrode to stimulate, and with what intensity, is transmitted across the skin using an inductive link 6 operating at radio frequencies, from an external multipeak speech processor (MSP) 7. In normal operation, the MSP picks up acoustic stimuli from a microphone 8 conveniently worn, and extracts from the signal, information which is used to determine stimulation electrode, rate and amplitude.

Because each patient's response to electrical stimulation is different, it is necessary to configure each patient's MSP to his or her own requirements. Thus, the MSP has a random access memory (RAM) which is programmed to suit each patient.

The patient's response to electrical stimulation is tested some short time after implantation of the RSU 3, using the patient's MSP, and the results of

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these tests are used to set up the MSP for the
patient's own particular requirements. This is done
by connecting the MSP, via a connector and cables 9,
to a diagnostic programming interface unit (DPI) 10.
The DPI is itself connected via a cable and connector
11 to a general purpose computer referred to as a
diagnostic and programming unit (DPU) 12.

A pictorial representation of the system
used by the patient is shown in FIGS. 3, 3A and 3B.
The electrode array 20 is flexible and fits the shape
of the cochlea (FIGS. 1A and 1B) as it is inserted
along the basilar membrane separating the scala tympa-
ni from the remainder of the cochlea. The electrode
array is connected via a silastic-covered cable 21 to
the RSU 22. Cable 21 is specially designed to provide
stress relief to prevent fracture of the wire in the
cable. The receiving coil for information and power
is a single turn of multistrand platinum wire 23 which
is transformer coupled to the implanted electronics in
the RSU 22.

An externally worn transmitting coil 24 is
held against the head over the site of the RSU implant
22 by, for example, cooperating magnets (not shown)
carried adjacent each of the coils 23 and 24, or by a
fixture (not shown) attached to the coil 24 for hold-
ing the coil to the user's head, or by adhesive tape.
Coil 24 is connected to the speech processor 29 via a
coil cable 26 and a hearing aid microphone 27. Hear-
ing aid microphone 27 is worn on the ear nearest to
the implant site and audio data from the microphone 27
is connected via a three wire cable 28 to the MSP 29.
Transmission data is connected to the coil 24 from the
MSP 29 via the same three wire cable 28 and via the
coil cable 26. This three wire arrangement is
described in the copending U.S. Patent Application No.

404,230, filed September 7, 1989, of Christopher N.
Daly, entitled "Three Wire System For Cochlear Implant
Processor," which application is assigned to the as-
signee of the present invention and is incorporated
herein by reference. Alternative microphone configu-
rations are possible, including a microphone worn on a
tie clasp or attached to the users clothing, or the
like.

The coil cable 26 and three wire cable 28
are attached to the microphone 27 and MSP 29 by de-
mountable connectors 32, 33 and 34. The MSP 29 is
powered by conventionally available batteries (e.g., a
single AA size cell) carried inside the MSP 29. A
plug-in jack 31 is provided to allow connection of
external audio signal sources, such as from a televi-
sion, radio, or high quality microphone.

Referring to FIG. 4, the pulse which is used
to electrically stimulate the cochlea is biphasic.
That is, it comprises a period of negative current
stimulation, followed by an equal period of positive
current stimulation of equal amplitude, the two peri-
ods (known as phases phi 1 and phi 2), separated by a
short period of no stimulation. Phi 1 and phi 2 may
be in the range of 12 to 400 microseconds (typically
200 microseconds), and the intervening interval is
typically about 50 microseconds. The amplitudes of
phi 1 and phi 2, their durations, and the duration of
the intervening interval are determined by the infor-
mation decoded from the signal transmitted by the
speech processor 29 (FIG. 3). The actual values of
these parameters will be set up on an electrode by
electrode basis, for each patient, as a result of
psychophysical testing of the patient. The reversal
in polarity of phi 1 and phi 2 is important since it
ensures that there is no net DC component in the stim-

ulus. This is important because long term DC excita-
tion might cause electrode corrosion, and possible
subsequent damage to the cochlea itself.

The questions of electrode electrochemistry
and charge balance are thought to be more important in
cochlear implants than in, say, cardiac pacemakers
which are well known in the art. This is because a
cochlear stimulator will be stimulating nerve fibers,
whereas a cardiac pacemaker is designed to stimulate
cardiac muscle. It is thought that nerve tissues may
be more susceptible to damage due to electrical stimu-
lation, and thus the cochlear implant system is de-
signed with more stringent safety factors than cardiac
pacemakers. The system is designed so that the same
stimulus source is used for both stimulation phases.
The biphasic pulse is produced simply by reversal of
the connections to the electrodes. Thus, extremely
good charge symmetry is obtained, resulting in a high
level of safety, provided the durations of phi 1 and
phi 2 are equal.

The stimulation circuitry is preferably
configured as a constant current source. This has the
advantage compared to a constant voltage source that
if the electrode impedance changes (as has often been
observed) the delivered current to the electrode will
remain unaltered over a large range of electrode im-
pedances. The current may be varied from a few micro-
amps to 2 mA, allowing a very large range of loudness
percepts to be produced and large variations between
patients to be accommodated.

The stimulus generation circuitry in the RSU
3 (FIG 2) is preferably designed to operate in one of
two modes. The first mode is referred to as "multipo-
lar" or "common ground" stimulation. In this mode,
one electrode is selected to be the "active" elec-

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trode, and all other electrodes operate as a common
current source. In phase phi 2, the connections are
reversed so that the "active" electrode acts as the
current source and the common electrodes act as a
current sink. The choice of stimulus order is not
determined by any limitations or restrictions in the
circuit design, and either way may be chosen when
implementing the circuit design.

The second mode is "bipolar" stimulation.
In this mode stimulation is between two selected elec-
trodes, let us say A and B. In phase phi 1, current
is sourced by A, and sunk by B. In phase phi 2, cur-
rent is sourced by B, and sunk by A, and no other
electrodes play any part in stimulation. The RSU 3 is
preferably configured so that any pair of electrodes
may be selected for bipolar stimulation. Thus there
is great flexibility in choice of stimulation strate-
gy.

It should be understood that only these two
particular stimulation modes have been chosen. Other
stimulation modes are not excluded, however. For
example, a multipolar or distributed ground system
could be used where not all other electrodes act as a
distributed ground, and any electrode could be
selected at any time to be a current source, current
sink, or inactive during either stimulation phase with
suitable modification of the receiver-stimulator.

The main aim of this invention is to provide
improved speech communication to those people suffer-
ing from profound hearing loss. However, in addition
to providing improved speech communication, it is also
important to be able to convey environmental sounds,
for example telephones, doors, warning sirens, door
bells, etc., which form part of a person's life. The
system described up to now is basically that of the

Crosby et al. patent, heretofore referred to and in-
corporated herein by reference. In the Crosby et al.
patent it is recognized that the second formant F2
carries most of the intelligibility of the speech
signal, while the first formant F1, although contain-
ing much of the naturalness of the signal, contributes
little to intelligibility.

     Crosby et al. observed that the third and
higher formants do not carry as much information as
the second formant. They also felt that in view of
the then limitations of knowledge on the interaction
between electrodes when a number of electrodes are
stimulated simultaneously, the most effective method
of stimulation would be to code the second formant on
an appropriate electrode or site in the cochlea to
provide the most important formant information. The
amplitude of such stimulation is derived from the
amplitude of the second formant.

     The Crosby et al. system also provides pro-
sodic information in the form of pulse rate. That
system compresses the stimulation rate to the range
100 - 250 Hz, the range in which the greatest pitch
discrimination from stimulation pulse rate is
achieved.

     An additional factor employed in Crosby et
al. is that only the top 10 to 20 dB of current acous-
tic stimulus level is used to determine stimulus am-
plitude. That is, instead of compressing the entire
acoustic loudness range into the small range of elec-
trical stimulation available, only the top part is
used. Thus, Crosby et al.'s amplitude of the signal
is entirely represented by a five bit binary code,
which provides only 30 dB of dynamic range.

     In summary, the Crosby et al. speech pro-
cessing strategy is:

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1.   The dominant spectral peak in the range
of about 300 Hz to about 4000 Hz is used to encode
electrode position.

2.   The amplitude of the dominant spectral
peak used to encode electrode position is used to
determine stimulation amplitude.

3.   Voice pitch (F0) is compressed and used
to determine the stimulation rate.

For unvoiced sounds and, environmental
sounds, the Crosby et al. system still generates stim-.
uli, but the stimulation rate and electrode position
will be determined by the exact nature of the acoustic
signal. For example, for sibilant consonants ("s"),
the stimulation rate is fairly fast, but not constant,
and the electrode stimulated will be one which illi-
cits a high frequency percept.

A second speech processing strategy, useful
in some patients, is employed in Crosby et al. The
second strategy is similar to the one mentioned above
in that electrode position is encoded from formant
frequency. However, the stimulation rate is at the F1
of first formant frequency, and the stimulation ampli-
tude is determined for the value of the peak of the
acoustic signal at the time of the F1 peak. This has
the advantage that the stimulation rate is faster, and
elicits more natural sounding speech perceptions in
some patients. In addition, since the F1 signal is
amplitude modulated and temporally better than the F0
rate, the patients also perceive the F0 or voice pitch
which is useful for conveying prosodic information.

Another speech processing strategy consid-
ered in the Crosby et al. reference is to stimulate
the patient at the rate of F1 extracted from an incom-
ing speech signal, but to pattern the stimulation such
that the stimuli are gated at the F0 rate.

Notwithstanding the success of speech pro-
cessors using the Crosby et al. F0, F1, F2 speech
processing coding scheme over the last few years, a
number of problems still remain in connection with the
use of such speech processor coding schemes. As indi-
cated earlier, patients who perform well in quiet con-
ditions can have significant problems when there is a
moderate level of background noise. Moreover, since
the F0,F1,F2 scheme codes frequencies up to about
4000Hz, and many phonemes and environmental sounds
have a high proportion of their energy above this
range, such phonemes and environmental sounds are
inaudible to the implant user in some cases.

In accordance with the present invention
multichannel cochlear implant prostheses having a
pulsatile operating system, such as that disclosed in
the Crosby et al. reference, are provided with a
speech coding scheme in which the speech signal is
bandpass filtered into a number of bands, for example
3, within and beyond the normal range of the second
frequency peak or formant F2 of the speech signal.
The speech coding scheme disclosed herein is referred
to as the multi-spectral peak coding strategy (MPEAK).
MPEAK is designed to provide additional high-frequency
information to aid in the perception of speech and
environmental sounds.

The MPEAK coding strategy extracts and codes
the F1 and F2 spectral peaks, using the extracted
frequency estimates to select a more apical and a more
basal pair of electrodes for stimulation. Each se-
lected electrode is stimulated at a pulse rate equal
to the fundamental frequency F0. In addition to F1
and F2, three high frequency bands of spectral infor-
mation are extracted. The amplitude estimates from
band three (2000-2800 Hz), band four (2800-4000 Hz),

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and band five (above 4000 Hz) are presented to fixed
electrodes, for example the seventh, fourth and first
electrodes, respectively, of the electrode array 1
(FIG. 2).

The first, fourth and seventh electrodes are
selected as the default electrodes for the high-fre-
quency bands because they are spaced far enough apart
so that most patients will be able to discriminate
between stimulation at these three locations. Note
that these default assignments may be reprogrammed as
required. If the three high frequency bands were
assigned only to the three most basal electrodes in
the MAP, many patients might not find the additional
high frequency information as useful since patients
often do not demonstrate good place-pitch discrimina-
tion between adjacent basal electrodes. Additionally,
the overall pitch percept resulting from the electri-
cal stimulation might be too high.

Table I below indicates the frequency ranges
of the various formants employed in the speech coding
scheme of the present invention.

TABLE I

Frequency Range	Formant or Band
280 - 1000 Hz	F1
800 - 4000 Hz	F2
2000 - 2800 Hz	Band 3 - Electrode 7
2800 - 4000 Hz	Band 4 - Electrode 1
4000 Hz and above	Band 5 - Electrode 1

If the input signal is voiced, it has a
periodic fundamental frequency. The electrode pairs
selected from the estimates of F1, F2 and bands 3 and
4 are stimulated sequentially at the rate equal to F0.
The most basal electrode pair is stimulated first,
followed by progressively more apical electrode pairs,
as shown in FIG. 5. Band 5 is not presented in FIG. 5

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because negligible information is contained in this
frequency band for voiced sounds.

        If the input energy is unvoiced, energy in
the F1 band (280-1000 Hz) is effectively zero. Conse-
quently it is replaced with the frequency band that
extracts information above 4000 Hz. In this situa-
tion, the electrodes pairs selected from the estimates
of F2, and bands 3, 4 and 5 receive the pulsatile
stimulation. The rate of stimulation is aperiodic and
varies between 200-300 Hz. FIG. 6 shows the sequen-
tial stimulation pattern for an unvoiced sound, with
stimulation progressing from base to apex. The MPEAK
coding strategy thus may be seen to extract and code
five spectral peaks but only four spectral peaks are
encoded for any one stimulus sequence.

        FIG. 7 illustrates the pattern of electrical
stimulation for various steady state phonemes when
using the MPEAK coding strategy. A primary function
of the MAP is to translate the frequency of the domi-
nant spectral peaks (F1 and F2) to electrode selec-
tion. To perform this function, the electrodes are
numbered sequentially starting at the round window of
the cochlea. Electrode 1 is the most basal electrode
and electrode 22 is the most apical in the electrode
array. Stimulation of different electrodes normally
results in pitch perceptions that reflect the tono-
topic organization of the cochlea. Electrode 22 elic-
its the lowest place-pitch percept, or the "dullest"
sound. Electrode 1 elicits the highest place-pitch
percept, or "sharpest" sound.

        To allocate the frequency range for the F1
and F2 spectral peaks to the total number of elec-
trodes, a default mapping algorithm splits up the
total number of electrodes available to use into a
ratio of approximately 1:2, as shown in FIG. 7. Con-

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sequently, approximately one third of the electrodes
are assigned to the F1 frequency range. These are the
more apical electrodes and they will cover the fre-
quency range of 280-1000 Hz. The remaining two thirds
of the electrodes are assigned to the F2 frequency
range (800-4000 Hz). The most apical electrodes,
which cover the frequency range from 280-1000 Hz, are
assigned linearly equal frequency bands. The frequen-
cy range corresponding to the estimate of F2 is as-
signed to the remaining more basal electrodes and is
divided into logarithmically equal frequency bands.
This frequency distribution is called linear/log
(lin/log) spacing.

A second optional mapping algorithm (not
shown) splits up the total frequency range into logar-
ythmically equal frequency bands for both F1 and F2
electrode groups (log/log spacing). In comparison to
the lin/log spacing, this results in relatively broad
frequency bands for electrodes that are assigned fre-
quency boundaries below 1000 Hz. Because of the wider
frequency bands for these electrodes, many vowel
sounds will stimulate similar electrodes, thus making
discrimination of these vowels difficult.

The F1/F2 lin/log function of the default
algorithm is preferable because it gives better spa-
tial resolution in the F1 range than the log/log func-
tion. In addition, this algorithm provides discrimi-
nation of vowels and consonants with formants close to
1000 Hz.

The mapping section of the DPS program al-
lows flexibility in assigning frequency bands to elec-
trodes. If fewer electrodes are included in the MAP,
then fewer and wider frequency bands are allocated
automatically by the computer so that the entire fre-
quency range is covered. Furthermore, it is possible

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to override the computer-generated spacing of frequen-
cy bands. Any range of frequencies may be allocated
to any electrode or electrodes by changing the upper
frequency boundaries.

　　　　Table II, below, shows the default bound-
aries (lin/log) for a MAP created in the biphasic +1
mode using 20 electrode pairs and the MPEAK coding
strategy.

　　TABLE II - Lin/Log Frequency Boundaries for 20
　　　　　　　　Electrodes in a BP +1 Mode. Also
　　　　　　　　Shown are the electrode allocations
　　　　　　　　for the three high frequency bands.

	Frequency Boundaries	
Electrode	Lower	Upper
20	280	400
19	400	500
18	500	600
17	600	700
16	700	800
15	800	900
14	900	1000
13	1000	1112
12	1112	1237
11	1237	1377
10	1377	1531
9	1531	1704
8	1704	1896
7	1896	2109
6	2109	2346
5	2346	2611
4	2611	2904
3	2904	3231
2	3231	3595
1	3595 & above	

　　　　　　　Electrodes: for Band 3 - 7
　　　　　　　　　　　　　　for Band 4 - 4
　　　　　　　　　　　　　　for Band 5 - 1

　　　　Table III, below, shows the default bound-
aries in the same mode using only 14 electrode pairs
and the MPEAK coding strategy.

　　TABLE III - Lin/Log Frequency Boundaries for 14
　　　　　　　　 Electrodes in a BP +1 Mode. Also

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shown are the electrode allocations
for the three high frequency bands.

| | Frequency Boundaries | |
Electrode	Lower	Upper
20	280	400
18	400	550
17	550	700
16	700	850
15	850	1000
14	1000	1166
13	1166	1360
10	1360	1587
9	1587	1851
8	1851	2160
7	2160	2519
6	2519	2939
5	2939	3428
4	3428 & above	

Electrodes: for Band 3 - 8
             for Band 4 - 6
             for Band 5 - 4

The amplitude of the electrical stimulus is
determined from the amplitude of the incoming acoustic
signal within each of the five frequency bands (F1,
F2, Bands 3, 4 and 5). However, because the elec-
trodes have different threshold (T) and maximum ac-
ceptable loudness (C) levels, the speech processor
must determine the level of stimulation for each elec-
trode separately based on the amplitude of the incom-
ing signal in each band.

The MSP (FIG. 2) contains a non-linear loud-
ness growth algorithm that converts acoustic signal
amplitude to electrical stimulation parameters.
First, the MSP converts the amplitude of the acoustic
signal into a digital linear scale with values from 0
to 150, as may be seen now by reference to FIG. 8.
That digital scale (in combination with the T and C-
levels stored in the patient's MAP) determines the
actual charge delivered to the electrodes. Signals

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whose amplitude levels are coded as 1 will cause stim-
ulation at the T-level. Signals whose amplitude lev-
els are coded as 150 will cause stimulation at the C-
level.

Referring now to FIG. 9, a block diagram of
the microphone and speech processor portions of a
pulsatile type, multi-channel cochlear implant system
100 have there been illustrated. The system 100 in-
cludes a microphone 110 which picks up speech and
provides electrical audio signals to a speech feature
extractor 112 through an automatic gain control ampli-
fier 111. The speech feature extractor 112 analyzes
the signals and provides digital outputs corresponding
to the frequencies and amplitudes of the first and
second formants, identified as F1, A1, F2 and A2,
respectively, in FIG. 10.

The speech feature extractor 112 also de-
tects and outputs the voice pitch FO and starts the
encoder 113, which translates, using a MAP 114 con-
taining information on the patient's psychophysical
test results, the voice pitch information into a pat-
tern of electrical stimulation on two electrodes that
are stimulated sequentially. The data so translated
is sent by the patient coil 115 to the implanted re-
ceiver stimulator unit RSU 3 (FIG. 2).

Three bandpass filters 116, 117 and 118 also
receive the audio signal from microphone 110 before it
is applied to the speech feature extractor 112, and
separate the signal into three components of different
frequencies, a 2000-2800 Hz signal in band 3, a 2800-
4000 Hz signal in band 4, and 4000-8000 Hz signal in
band 5. The signals from bands 3, 4 and 5 are lead to
the encoder 113 and mapping of these signals is done
in a manner similar to that for the first and second
formants, and translation of the resulting pattern of

electrical stimulation to the appropriate electrodes
takes place, as discussed earlier herein.

The automatic gain control amplifier 111 is
used to control the amplitude of the signal fed to the
filters 116 and 117. Since filter 118 is only used
for unvoiced parts of the speech signal, its amplitude
is never very great and, therefore, the signal does
not require automatic gain control. Accordingly,
amplifier 119 does not have automatic gain control
provisions incorporated therein.

To summarize, the psychophysical measure-
ments that are made using the DPS software provide the
information for translating the extracted acoustic
input into patient-specific stimulation parameters.
Threshold (T) and maximum (C) levels for electrical
stimulation are measured for each electrode pair.
These values are stored in the MAP. They determine
the relationship between the incoming acoustic signal
amplitude and the stimulation level for any given
electrode pair.

Inside the speech processor a random access
memory stores a set of number tables, referred to
collectively as a MAP. The MAP determines both stimu-
lus parameters for F1, F2 and bands 3-5, and the am-
plitude estimates. The encoding of the stimulus pa-
rameters follows a sequence of distinct steps. The
steps may be summarized as follows:

1. The first formant frequency (F1) is
converted to a number based on the dominant spectral
peak in the region between 280-1000 Hz.

2. The F1 number is used, in conjunction
with one of the MAP tables, to determine the electrode
to be stimulated to represent the first formant. The
indifferent electrode is determined by the mode.

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3. The second formant frequency (F2) is
converted to a number based on the dominant spectral
peak in region between 800-4000 Hz.

4. The F2 number is used, in conjunction
with one of the MAP tables to determine the electrode
to be stimulated to represent the second formant. The
indifferent electrode is determined by the mode.

5. The amplitude estimates for bands 3, 4
and 5 are assigned to the three default electrodes 7,
4 and 1 for bands 3, 4 and 5, respectively, or such
other electrodes that may be selected when the MAP is
being prepared.

6. The amplitude of the acoustic signal in
each of the frequency bands is converted to a number
ranging from 0 - 150. The level of stimulation that
will be delivered is determined by referring to a set
MAP tables that relate acoustic amplitude (in range of
0-150) to stimulation level for the specific elec-
trodes selected in steps 2, 4 and 5, above.

7. The data are further encoded in the
speech processor and transmitted to the receiver/stim-
ulator. It, in turn, encodes the data and sends the
stimuli to the appropriate electrodes. Stimulus
pulses are presented at a rate equal to F0 during
voiced periods and at a random aperiodic rate within
the range of F0 and F1 formants (typically 200 to
300 Hz) during unvoiced periods.

It will be apparent from the foregoing de-
scription that the multi-spectral peak speech coding
scheme of the present invention provides all of the
information available in the prior art F0F1F2 scheme,
while providing additional information from three high
frequency band pass filters. These filters cover the
following frequency ranges: 2000 to 2800 Hz, 2800 to
4000 Hz and 4000 to 8000 Hz. The energy within these

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ranges controls the amplitude of electrical stimula-
tion of three fixed electrode pairs in the basal end
of the electrode array. Thus, additional information
about high frequency sounds is presented at a tono-
topically appropriate place within the cochlea.

The overall stimulation rate remains as F0
(fundamental frequency or voice pitch) but in the
scheme of the present invention four electrical stimu-
lation pulses occur for each glottal pulse. This
compares with the prior F0F1F2 strategy in which only
two pulses occur per voice pitch period. In the new
coding scheme, for voiced speech sounds, the two
pulses representing the first and second formant are
still provided, and additional stimulation pulses
occur representing energy in the 2000-2800 Hz and the
2800-4000 Hz ranges.

For unvoiced phonemes, yet another pulse
representing energy above 4000 Hz is provided while no
stimulation for the first formant is provided, since
there is no energy in this frequency range. Stimula-
tion occurs at a random pulse rate of approximately
260 Hz, which is about double that used in the earlier
strategy.

It will be further apparent from the forego-
ing description that this invention provides an im-
proved cochlear implant system which overcomes various
of the problems associated with earlier cochlear im-
plant systems. The use of a multi-spectral peak
speech coding strategy in accordance with this inven-
tion provides the user of the implant system with
significantly improved speech recognition, even in the
presence of moderate levels of background noise. In
addition improved recognition of phonemes and environ-
mental sounds are provided by this invention.

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        While a particular embodiment of this inven-
tion has been shown and described, it will be obvious
to those skilled in the art that various changes and
modifications may be without departing from this in-
vention in its broader aspects, and it is, therefore,
aimed in the appended claims to cover all such changes
and modifications as fall within the true spirit and
scope of this invention.

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Claims

What is claimed is:

1. A multi-channel cochlear prosthesis including a patient implantable tissue stimulating multi-channel electrode array adapted to be positioned in a cochlea from the apical region of the cochlea to the basal region of the cochlea, a patient implantable multi-channel stimulator connected to said array, and a patient externally worn programmable speech processor for processing sound signals into electrical stimulation signals that are transmitted to said stimulator, said prosthesis further comprising:

means based on a dominant peak extraction in the region of between about 280 Hz to about 1000 Hz for determining a first formant spectral information in said sound signal and stimulating at least one electrode in the apical region of said electrode array in accordance with the spectral information of said formant;

means based on a dominant spectral peak extraction in the region of between about 800 Hz and about 4000 Hz for determining a second formant frequency in said sound signal and stimulating at least one electrode in said basal region of said electrode array in accordance with the spectral information of said formant; and,

at least one high frequency band filter for extracting spectral information in at least one region of the spectrum of said sound signal and stimulating at least one predetermined electrode in said electrode array in accordance with said extracted spectral information, said predetermined electrode being in said basal region of said electrode array.

2. A multi-channel cochlear prosthesis according to claim 1, including a plurality of said

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high frequency band filters for extracting spectral
information in a corresponding number of regions of said
sound signal and stimulating at least a corresponding
number of said predetermined electrodes in said elec-
trode array, all of said predetermined electrodes being
in said basal region of said electrode array.

3.   A multi-channel cochlear prosthesis
according to claim 2, wherein said electrical stimula-
tion signals are applied to said electrodes in the
form of pulses presented at a pulse rate dependent on
the pitch of the sound signal.

4.   A multi-channel cochlear prosthesis
according to claim 3, wherein said pulse rate is in
the range of between about 80 Hz and about 400 Hz.

5.   A multi-channel cochlear prosthesis
according to any one of claims 1 - 4 including at
least three of said high frequency band filters each
with predetermined electrodes.

6.   A multi-channel cochlear prosthesis
according to claim 5, wherein a first one of said high
frequency band filters extracts spectral information
from the sound signal in a frequency range of between
about 2000 Hz and about 2800 Hz, wherein a second one
of said high frequency band filters extracts spectral
information from the sound signal in a frequency range
of between about 2000 Hz and about 2800 Hz, and where-
in a third one of said high frequency band filters
extracts spectral information from the sound signal in
a frequency range of between about 4000 Hz and about
8000 Hz.

7.   A multi-channel cochlear prosthesis
according to claim 6, wherein the electrodes in said
electrode array may be considered to be consecutively
numbered, starting from the basal end thereof and
extending to the apical end thereof, and wherein am-

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plitude estimates derived from the spectral informa-
tion extracted from said first, second and third high
frequency band filters are applied to said correspond-
ing number of predetermined electrodes with the ampli-
tude estimate from said first filter being applied to
a higher-numbered electrode than the amplitude esti-
mate from said second filter and the amplitude esti-
mate from said second filter being applied to a
higher-numbered electrode than the amplitude estimate
from said third filter.

        8.   A multi-channel cochlear prosthesis
according to claim 7, wherein said electrode array
includes about 22 electrodes therein, wherein said
basal region of said electrode array comprises about
two-thirds of the electrodes in said electrode array,
and wherein said apical region comprises about one-
third of the electrodes in said electrode array.

        9.   A multi-channel cochlear prosthesis
according to claim 8, wherein said amplitude estimates
derived from the spectral information extracted from
said first, second and third high frequency band fil-
ters are applied to said seventh, fourth and first
electrodes, respectively, in said electrode array.

        10. A multi-channel cochlear prosthesis
according to claim 7, wherein, in the case of voiced
sound signals, the electrodes selected to be stimu-
lated are based on the first and second formants and
on information derived from the first and second fil-
ters, and wherein said electrodes are stimulated se-
quentially at a rate that is based on the pitch of the
sound signal, with the most basal of said electrodes
being stimulated first, followed by stimulation of
progressively more apical electrodes.

        11. A multi-channel cochlear prosthesis
according to claim 7, wherein, in the case of unvoiced

sound signals, the electrodes selected to be stimulated are based on the second formant and on information derived from the first, second and third filters, and wherein said electrodes are stimulated sequentially at an aperiodic rate within the range of from formant F0 to formant F1, with the most basal of said electrodes being stimulated first, followed by stimulation of progressively more apical electrodes.

12.   A multi-channel cochlear prosthesis according to claim 7, wherein said electrodes are stimulated at an aperiodic rate within the range of about 200Hz to about 300Hz.

13.   A multi-channel cochlear prosthesis according to claim 11, wherein said aperiodic rate is within the range of about 200Hz to about 300Hz.

14.   A method of processing an audio spectrum signal received from a microphone to produce signals for stimulating a patient implantable tissue stimulating multi-channel electrode array adapted to be positioned in a cochlea from the apical region of the cochlea to the basal region of the cochlea, said method comprising selecting a first dominant frequency peak from said audio signal from a frequency band of between about 280Hz and about 1,000Hz and stimulating at least one electrode in the apical region of said electrode array in accordance with the spectral information contained in said first peak; selecting a second dominant frequency peak from said audio signal from a frequency band of between
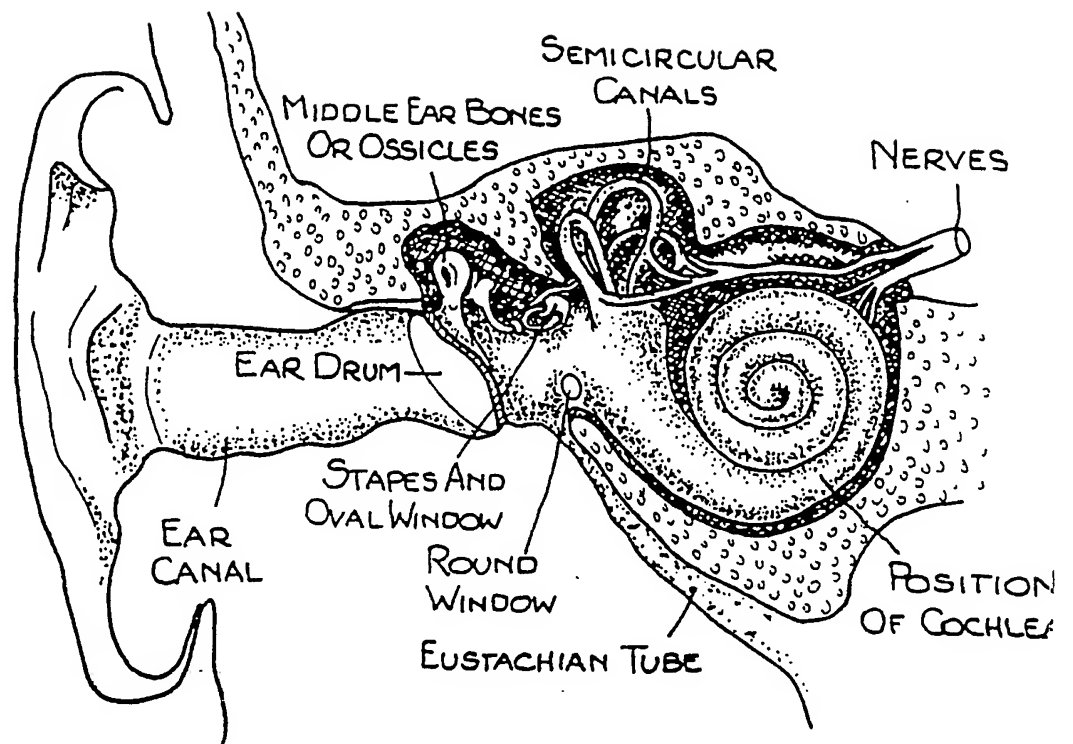
about 800Hz and about 4,000Hz and stimulating at least one
electrode in the basal region of said electrode array in
accordance with the spectral information contained in said
second peak; extracting spectral information in at least one
region of the spectrum of said audio signal and stimulating
at least one predetermined electrode in said electrode array
in accordance with said extracted spectral information, said
predetermined electrode being in said basal region of said
electrode array.

15.   A method of processing an audio spectrum signal as
claimed in claim 14 wherein additional preselected electrodes
are stimulated using spectral energy derived from said audio
signal in the audio frequency regions 2,000 to 2,800Hz, 2,800
to 4,000Hz and above 4,000Hz respectively.

AMENDED CLAIMS

[received by the International Bureau on
14 January 1991 (14.01.91);
new claims 16 and 17 added;
other claims unchanged (1 page)]

about 800Hz and about 4,000Hz and stimulating at least one electrode in the basal region of said electrode array in accordance with the spectral information contained in said second peak; extracting spectral information in at least one region of the spectrum of said audio signal and stimulating at least one predetermined electrode in said electrode array in accordance with said extracted spectral information, said predetermined electrode being in said basal region of said electrode array.

15.   A method of processing an audio spectrum signal as claimed in claim 14 wherein additional preselected electrodes are stimulated using spectral energy derived from said audio signal in the audio frequency regions 2,000 to 2,800Hz, 2,800 to 4,000Hz and above 4,000Hz respectively.

16.(new)   A method of speech coding an audio spectrum signal received from a microphone to produce signals for stimulating a patient implantable tissue stimulating multi-channel electrode array adapted to be positioned in a cochlea from the apical region of the cochlea to the basal region of the cochlea, said method comprising bandpass filtering said audio spectrum signal into a plurality of bands within and beyond the normal range of a second (F2) formant frequency peak of said audio spectrum signal whereby additional high frequency information is provided to a patient.
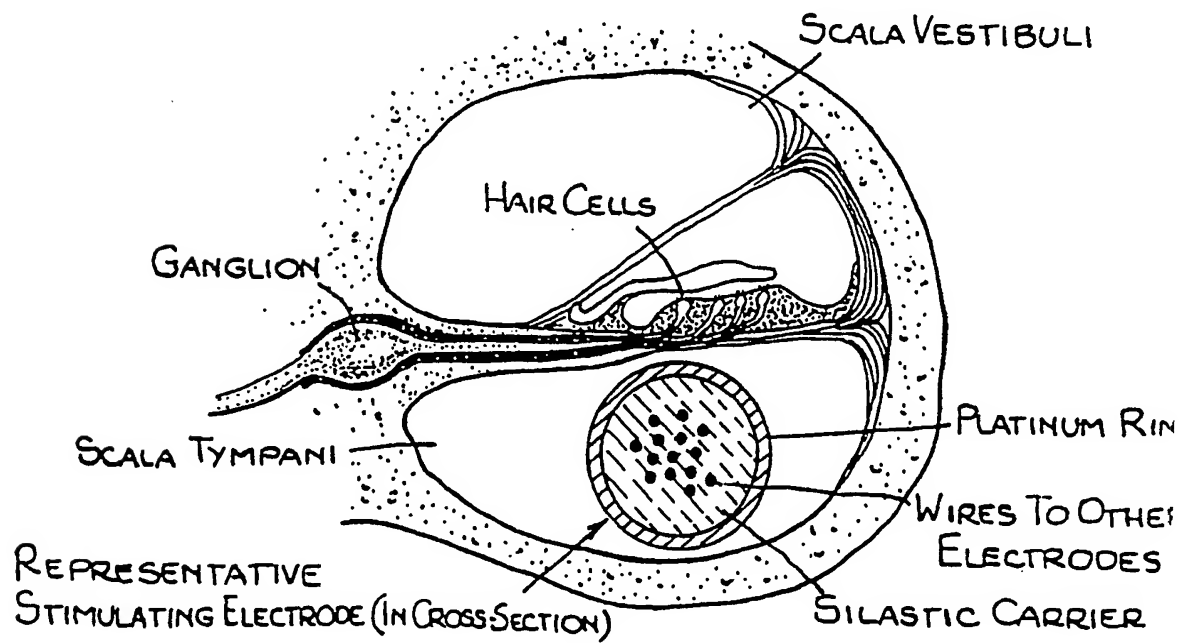
17.(new)   The method of claim 16 wherein information thus derived from said audio spectrum signal is encoded into sequential pulses and applied to selected electrodes of said electrode array.
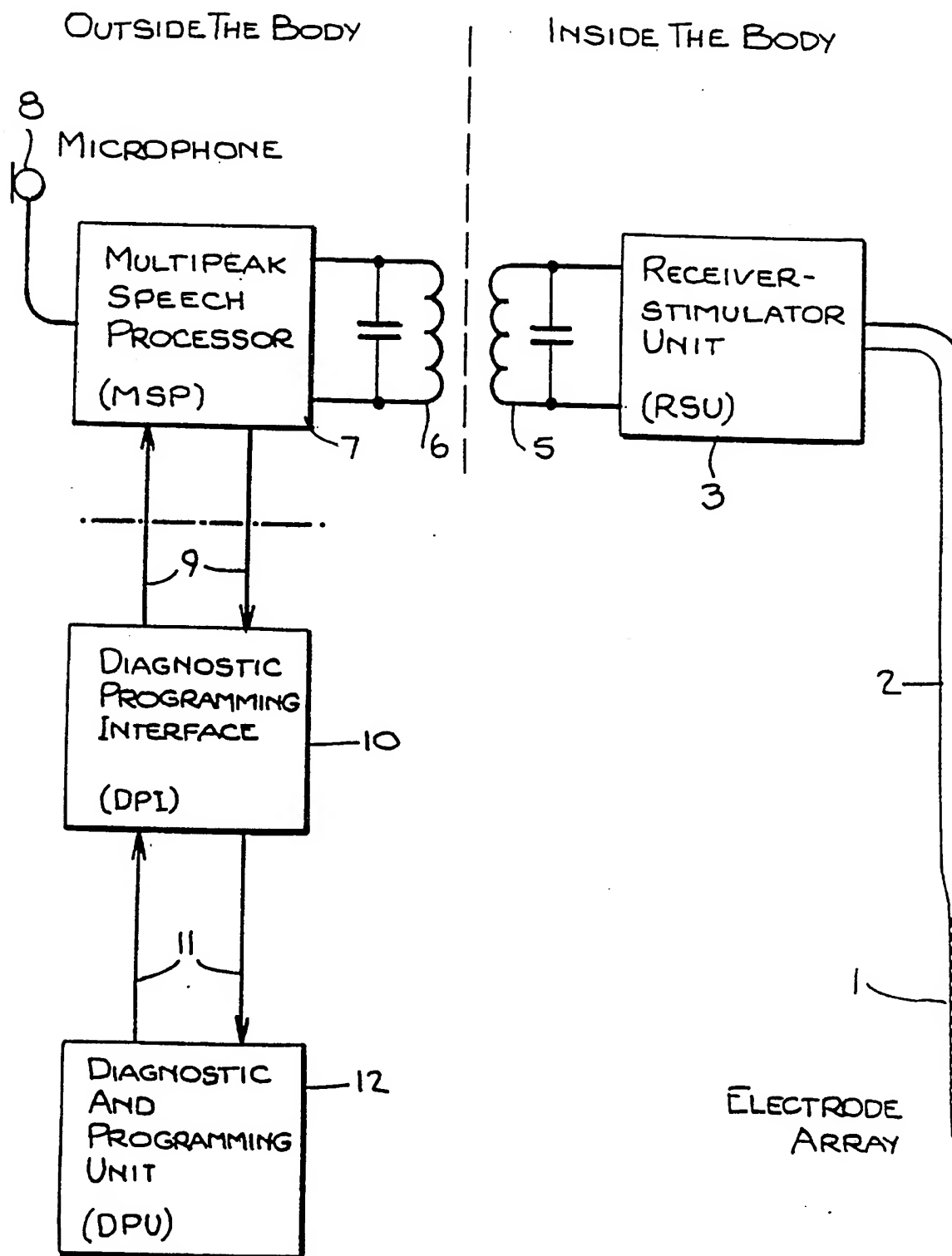
MIDDLE EAR BONES
OR OSSICLES

SEMICIRCULAR
CANALS

NERVES

EAR DRUM

STAPES AND
OVAL WINDOW

ROUND
WINDOW

EUSTACHIAN TUBE

EAR
CANAL

POSITION
OF COCHLEA

ANATOMICAL ARRANGEMENT

Fig.1A.

SCALA VESTIBULI

HAIR CELLS

GANGLION

SCALA TYMPANI

PLATINUM RIN

WIRES TO OTHER
ELECTRODES

SILASTIC CARRIER

REPRESENTATIVE
STIMULATING ELECTRODE (IN CROSS SECTION)

COCHLEA IN CROSS-SECTION

Fig.1B.
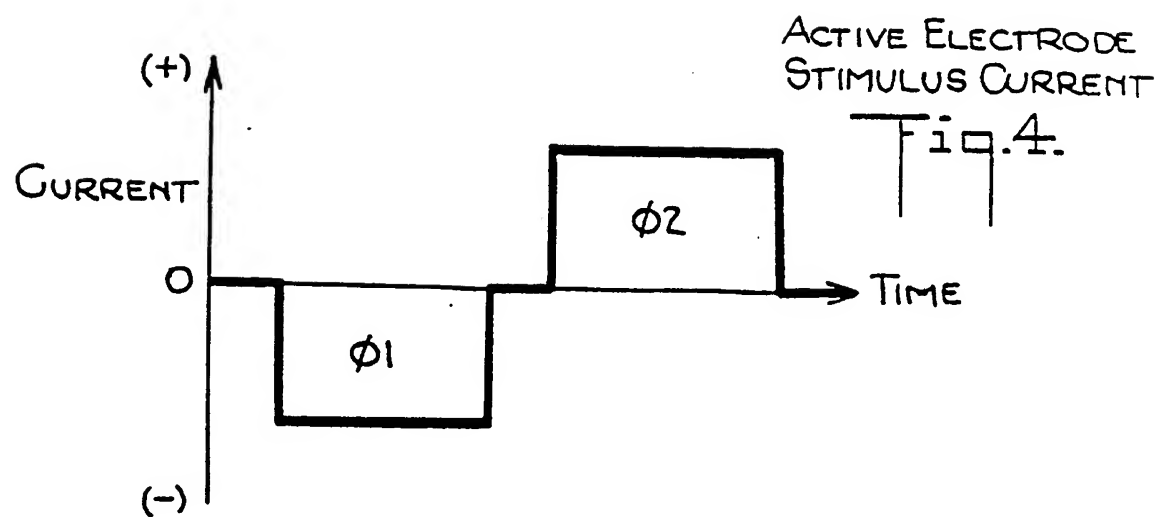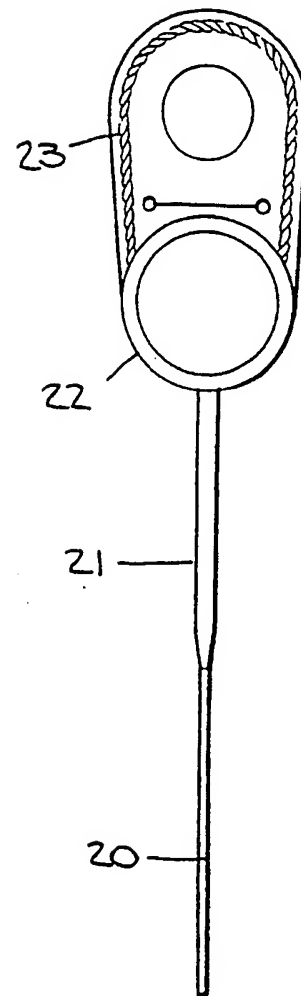
2/7   Fig.2.

COCHLEAR IMPLANT SYSTEM BLOCK DIAGRAM

OUTSIDE THE BODY   INSIDE THE BODY

8

MICROPHONE

MULTIPEAK
SPEECH
PROCESSOR

(MSP)

7   6

RECEIVER-
STIMULATOR
UNIT

(RSU)

5   3

9

DIAGNOSTIC
PROGRAMMING
INTERFACE   10

(DPI)

2

11

DIAGNOSTIC
AND
PROGRAMMING   12
UNIT

(DPU)

1

ELECTRODE
ARRAY

3/7

Fig.3.

Fig.3A.

Fig.3B.

Active Electrode
Stimulus Current

Fig.4.

FO PERIOD

BAND 5

BAND 4

BAND 3

F2 ELEC

F1 ELEC
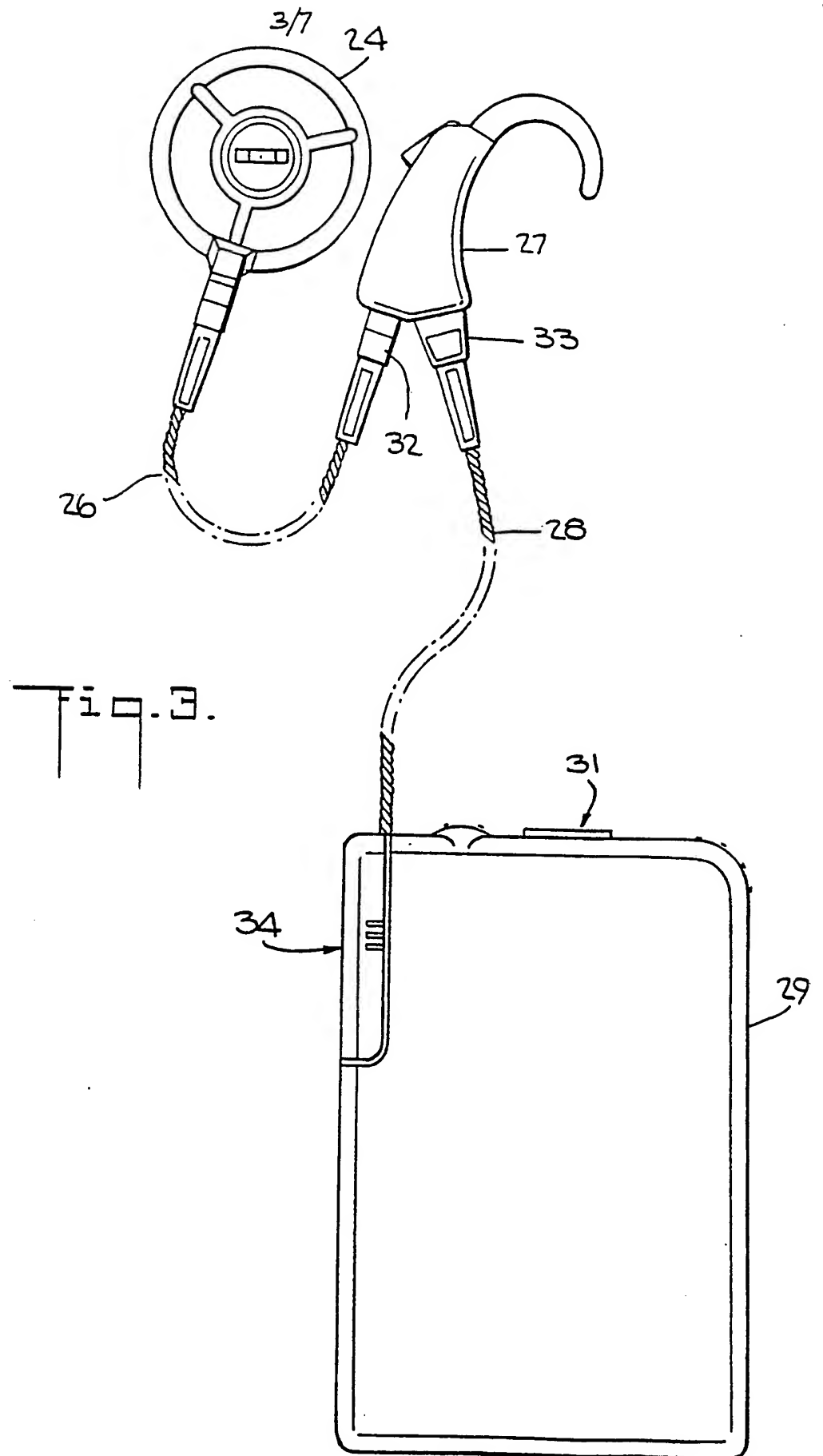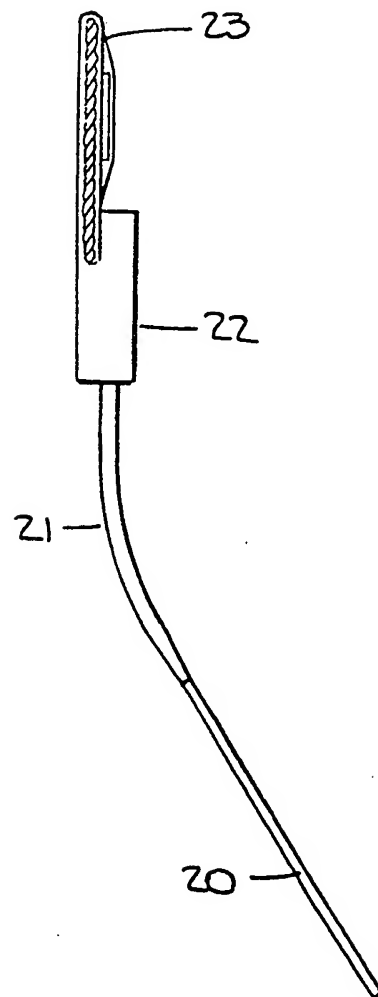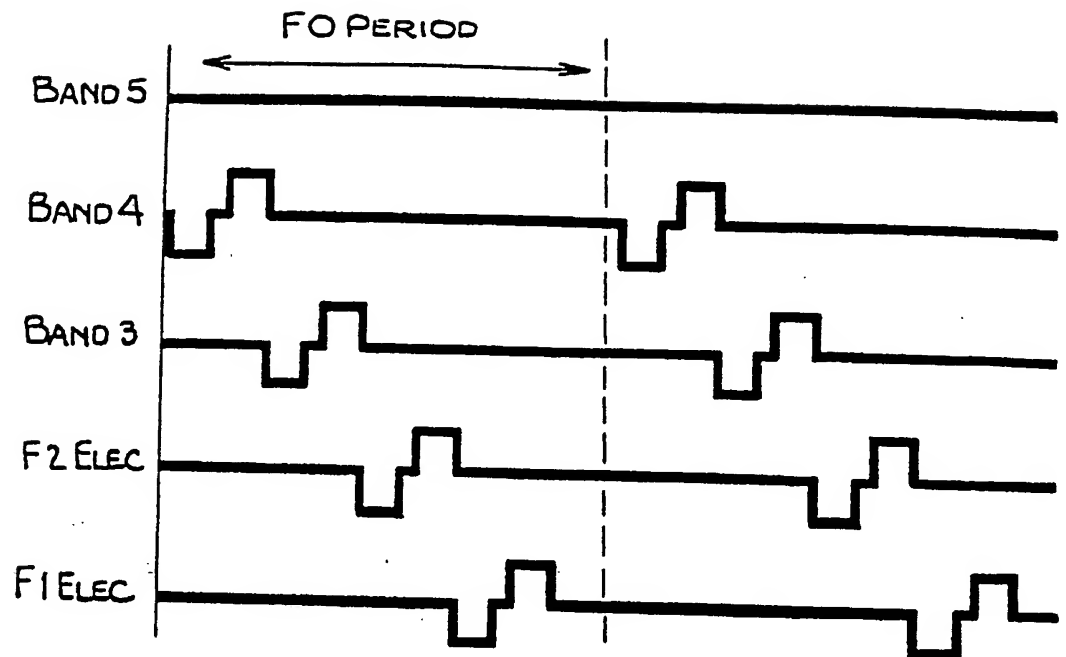
MPEAK STRATEGY-VOICED SOUNDS

Fig. 5.

PERIOD RANDOM AROUND 4MS

BAND 5

BAND 4

BAND 3

F2 ELEC

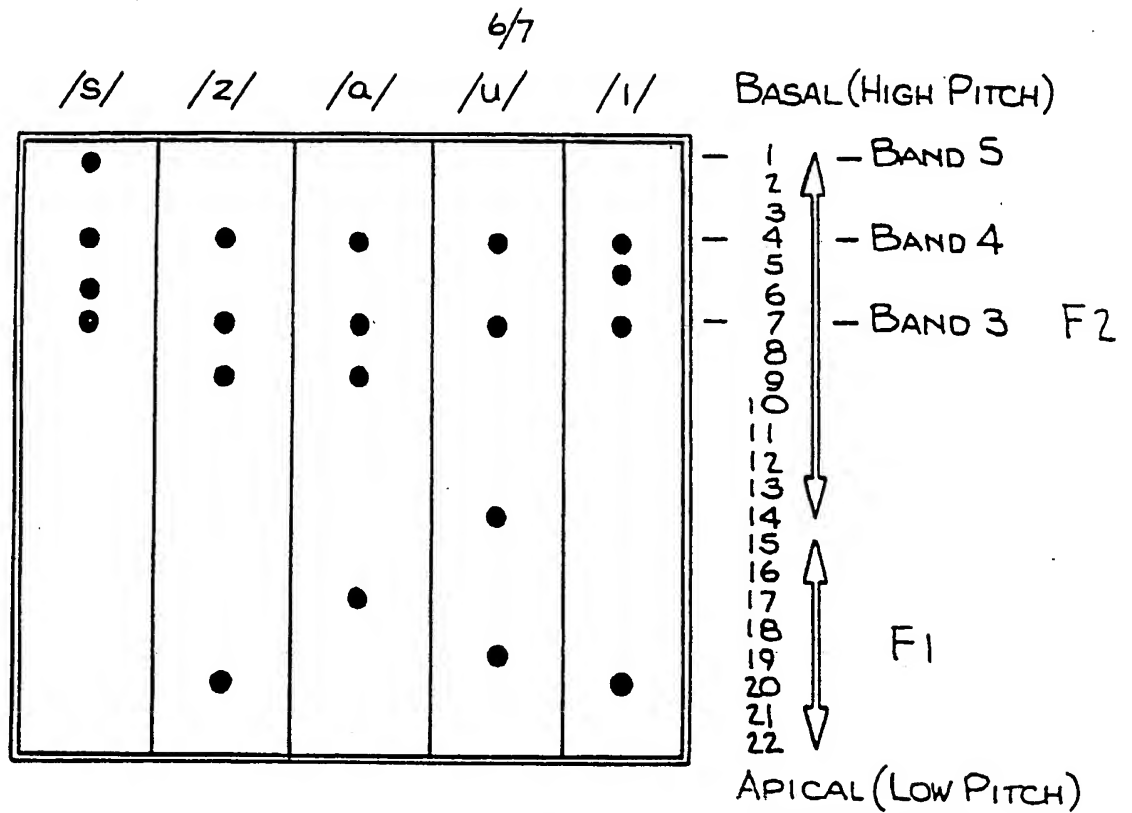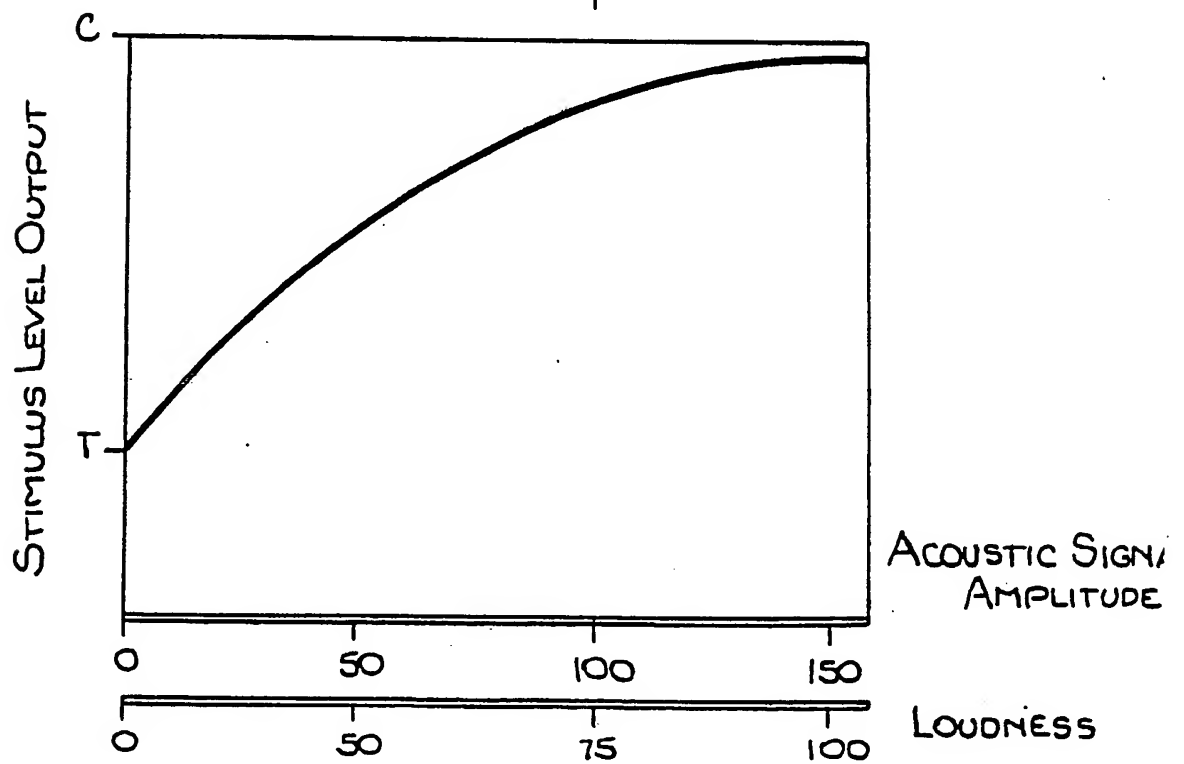F1 ELEC
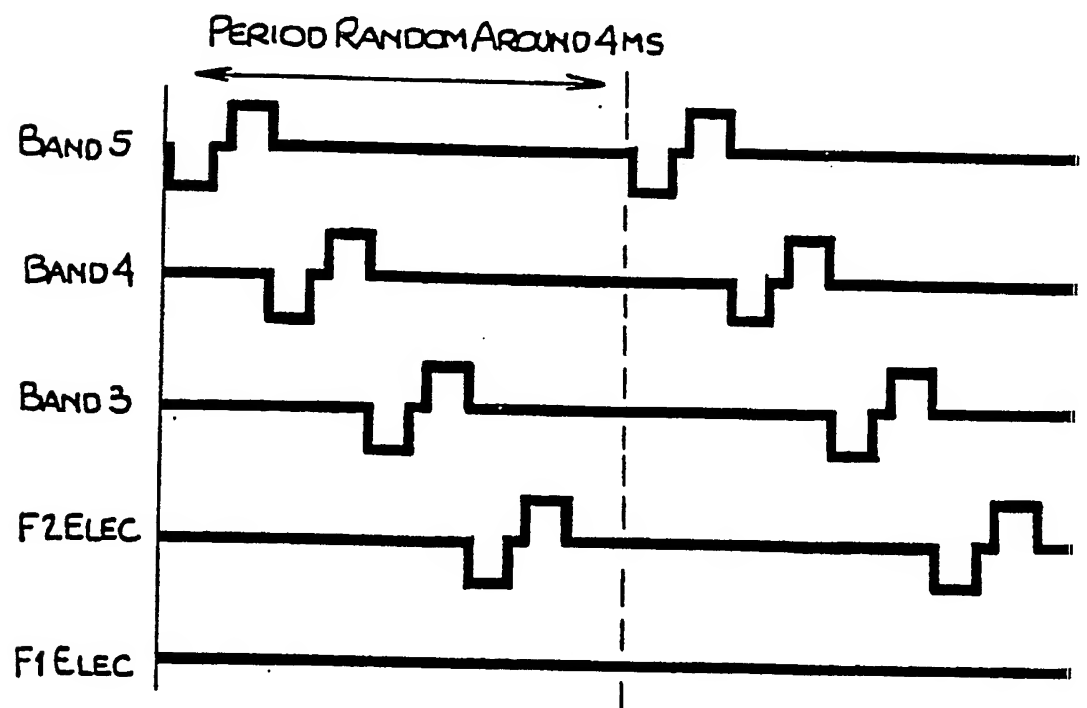
MPEAK STRATEGY-UNVOICED SOUNDS
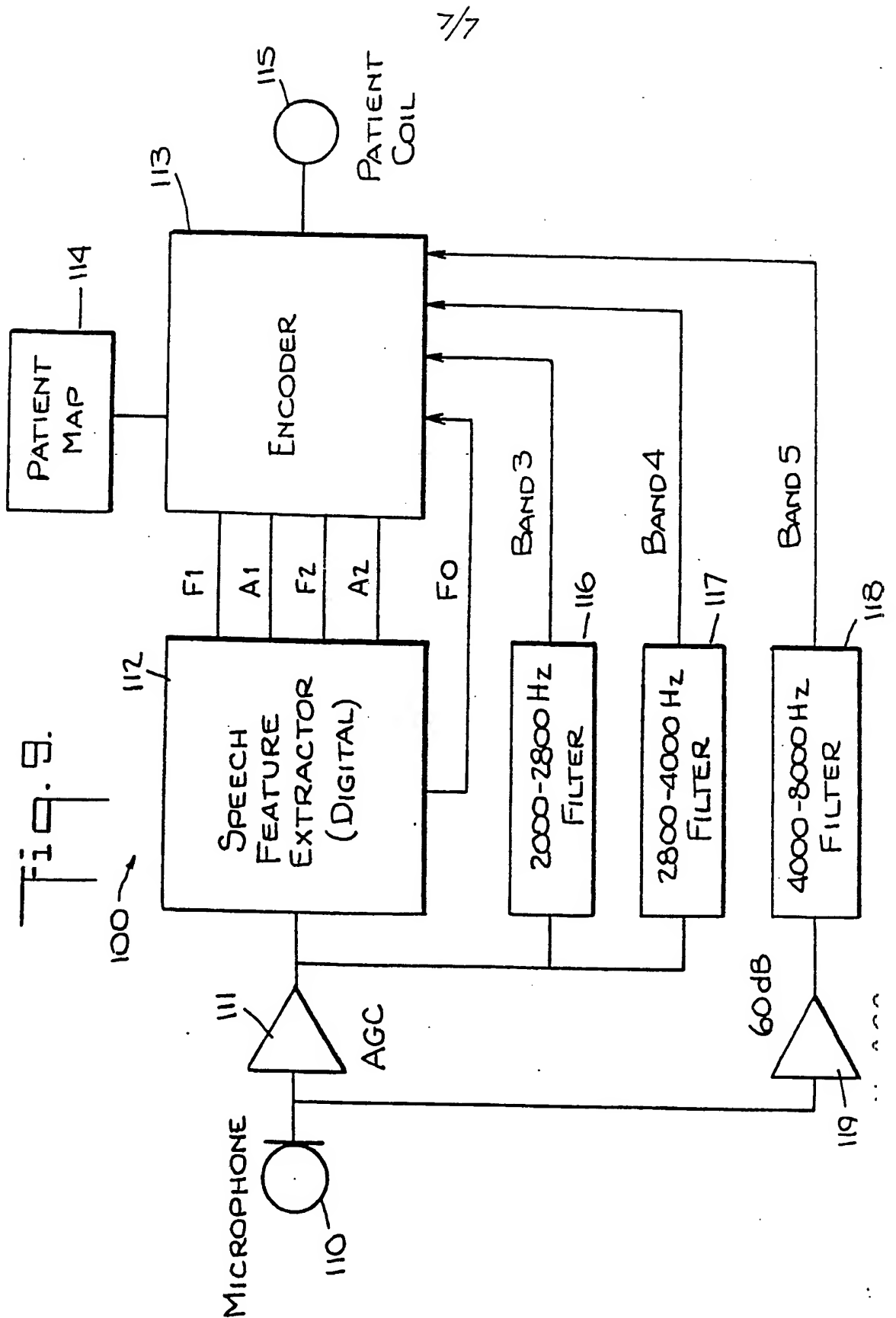
Fig. 6.

ELECTRICAL STIMULATION PATTERNS FOR THE MPEAK STRATEGY

Fig. 7.

STANDARD LOUDNESS GROWTH FUNCTION FOR THE WSP

Fig. 8.

7/7

Fig. 9.

I. CLASSIFICATION OF SUBJECT MATTER (if several classification symbols apply, indicate all) 6

According to International Patent Classification (IPC) or to both National Classification and IPC

Int. Cl.^{5} H04R 25/00, A61F 2/18

II. FIELDS SEARCHED

| Minimum Documentation Searched 7 |

Classification System	Classification Symbols
I..	H04R 25/00, A61F 2/18, 1/18

Documentation Searched other than Minimum Documentation
to the Extent that such Documents are Included in the Fields Searched 8

AU :   IPC as above

III. DOCUMENTS CONSIDERED TO BE RELEVANT 9

Category*	Citation of Document, with indication, where appropriate, of the relevant passages 12	Relevant to Claim No 13
A	EP,A, 124930 (CROSBY et al) 14 November 1984 (14.11.84) See page 11 lines 8-32 and page 12 lines 1-8	
A	AU,A, 18194/83 (THE UNIVERSITY OF MELBOURNE) 23 February 1984 (23.02.84) See page 3 lines 6-24	
A	AU,A, 46815/85 (THE UNIVERSITYT OF MELBOURNE) 13 March 1986 (13.03.86) See page 5 lines 15-29 and page 6 lines 1-24	
A	US,A, 4593696 (HOCHMAIR et al) 10 June 1986 (10.06.86) See abstract	
A	US,A, 4611598 (HORTMANN et al) 16 September 1986 (16.09.86) See abstract	
	(continued)	

* Special categories of cited documents: 10

"A" document defining the general state of the art which is not considered to be of particular relevance
"E" earlier document but published on or after the international filing date
"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)
"O" document referring to an oral disclosure, use, exhibition or other means
"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention
"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step.
"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.
"&" document member of the same patent family

IV. CERTIFICATION

Date of the Actual Completion of the International Search 15 November 1990 (15.11.90)	Date of Mailing of this International Search Report 21 November 1990
International Searching Authority Australian Patent Office	Signature of Authorized Officer R. MURRAY

Form PCT/ISA/210 (second sheet) (January 1985)

	FURTH⁻R INFORMATION CONTINUED FROM THE SECOND SHEET		
A	US,A, 4063048 (KISSIAH) 13 December 1977 (13.12.77) See abstract	
A	US,A, 4592359 (GALBRAITH) 3 June 1986 (03.06.86) See abstract	

V. [ ] OBSERVATIONS WHERE CERTAIN CLAIMS WERE FOUND UNSEARCHABLE 1

This international search report has not been established in respect of certain claims under Article 17(2)(a) for the following reasons:

 1.[ ] Claim numbers ..., because they relate to subject matter not required to be searched by this Authority, namely:

 2.[ ] Claim numbers , because they relate to parts of the international application that do not comply with the prescribed requirements to such an extent that no meaningful international search can be carried out, specifically:

 3.[ ] Claim numbers ..., because they are dependent claims and are not drafted in accordance with the second and third sentences of PCT Rule 6.4 (a):

VI. [ ] OBSERVATIONS WHERE UNITY OF INVENTION IS LACKING 2

This International Searching Authority found multiple inventions in this international application as follows:

1.[ ] As all required additional search fees were timely paid by the applicant, this international search report covers all searchable claims of the international application.

2.[ ] As only some of the required additional search fees were timely paid by the applicant, this international search report covers only those claims of the international application for which fees were paid, specifically claims:

3.[ ] No required additional search fees were timely paid by the applicant. Consequently, this international search report is restricted to the invention first mentioned in the claims; it is covered by claim numbers:

4. [ ] As all searchable claims could be searched without effort justifying an additional fee, the International Searching Authority did not invite payment of any additional fee.

Remark on Protest
[ ] The additional search fees were accompanied by applicant's protest.
[ ] No protest accompanied the payment of additional search fees.

Form PCT/ISA/210 (supplemental sheet (2)) (January 1985)

This Annex lists the known "A" publication level patent family members
relating to the patent documents cited in the above-mentioned
international search report. The Australian Patent Office is in no way
liable for these particulars which are merely given for the purpose of
information.

Patent Document Cited in Search Report	Patent Family Members		
EP 124930	AU 26683/84 CA 1212501 JP 61001200	AU 73227/87 EP 241101 US 4532930	AU 73251/87 EP 247649
AU 46815/85	US 4947844		
US 4592359	AU 55350/86 EP 198618	AU 54985/90 JP 61293475	CN 86103049
US 4611598	DE 3420244	EP 163137	
US 4593696	AU 51720/85 CN 86100219	BR 8600156 EP 190836	CA 1247227 JP 61170457

END OF ANNEX